IMAGE-WIDE ARTIFACTS REDUCTION CAUSED BY HIGH ATTENUATING OBJECTS IN CT DEPLOYING VOXEL TISSUE CLASS

DESCRIPTION

The present application relates to the diagnostic imaging arts. It finds particular application in computed tomography imaging of a subject that includes high density regions such as metal implants, dental fillings, and the like, and will be described with particular reference thereto. However, it also finds application in other types of
5   tomographic imaging such as single photon emission computed tomography (SPECT), positron emission tomography (PET), three-dimensional x-ray imaging, and the like.

In CT imaging, high absorbing objects such as metal bodies may cause significant artifacts, which may compromise the diagnostic value of the image. Metal artifacts arise when the imaged region of interest contains metal implants, dental fillings,
10   bullets, or other articles of high radiation absorption which prevent the x-rays from fully penetrating the subject. Projection line integrals passing through the regions of high density are so highly attenuated by the high density regions that data about other regions along the line integral are lost or overshadowed. This leads to substantial measurement errors. The filtered backprojection or other reconstruction process translates these
15   measurement errors into image artifacts, e.g. streaks which emanate from the high intensity region. The streaks dramatically deteriorate image quality and can obliterate structure of the region.

A previous method for correcting metal artifacts includes performing filtered backprojection to generate an uncorrected reconstructed image, identifying a region
20   of high density in the uncorrected reconstructed image, and replacing projections that pass through the high density region with synthetic projection data having reduced absorption attenuation values. The corrected projection data again undergoes filtered backprojection to produce a corrected reconstructed image.

This known method works well for certain imaging applications in which
25   there is a single, well-defined high density region surrounded by much lower density tissue. It does not work well, however, with a plurality of high density regions, or where there are medium density regions in addition to the high density region. For such composite imaging subjects, metal artifacts are reduced but remain very visible in the

corrected reconstructed image, especially between high density and medium density regions. In medical imaging applications, medium density regions typically correspond to bone while high density regions typically correspond to metal implants, dental fillings, operation clips (used in certain interventional computed tomography applications),

5   prosthesis devices, and the like. Hence, in medical computed tomography imaging, the region of interest commonly contains medium density regions.

There is a need for an automated technique that compensates for metal artifacts in the image yet is simple, cost effective and easy to implement. The present invention contemplates a method and apparatus that overcomes the aforementioned

10 limitations and others.

According to one aspect of the present application, a diagnostic imaging system which automatically corrects metal artifacts in an uncorrected tomographic image

15 caused by high attenuating objects is disclosed. A means clusters pixels of the uncorrected tomographic image. A means classifies pixels of the uncorrected reconstructed image into at least metal, bone, tissue, and air pixel classes to generate a classified image. A means replaces metal class pixels of the classified image with pixel values of another pixel class to generate a metal free classified image. A means forward projects the metal free

20 classified image to generate a model projection data. A means identifies corrupted regions of original projection data contributing to the pixels of the metal class. A means replaces the identified corrupted regions with corresponding regions of the model projection data to generate corrected projection data which is reconstructed by a reconstruction means into a corrected reconstructed image.

25 According to another aspect of the present application, a method for automatically correcting metal artifacts in an uncorrected tomographic image caused by high attenuating objects is disclosed. Pixels of the uncorrected tomographic image are clustered. Pixels of the uncorrected reconstructed image are classified into at least metal, bone, tissue, and air pixel classes to generate a classified image. Metal class pixels of the

30 classified image are replaced with pixel values of another pixel class to generate a metal free classified image. The metal free classified image is forward projected to generate a model projection data. Corrupted regions of original projection data contributing to the

pixels of the metal class are identified. The identified corrupted regions are replaced with corresponding regions of the model projection data to generate corrected projection data. The corrected projection data is reconstructed into a corrected reconstructed image.

One advantage of the present application resides in automatically compensating the image artifacts caused by high attenuation objects.

Another advantage resides in automatically correcting image artifacts in images including a metal or other high density region and a bone or other medium density region.

Another advantage resides in correct radiation therapy planning for the subjects containing highly attenuating objects.

Another advantage resides in applicability of the present application to both original sinogram data and otherwise available reconstructed images.

Numerous additional advantages and benefits will become apparent to those of ordinary skill in the art upon reading the following detailed description of the preferred embodiments.

The invention may take form in various components and arrangements of components, and in various process operations and arrangements of process operations. The drawings are only for the purpose of illustrating preferred embodiments and are not to be construed as limiting the invention.

FIGURE 1 diagrammatically shows a computed tomography imaging system including an artifact-correcting process according to the present application;

FIGURE 2 diagrammatically shows an expanded portion of the computed tomography imaging system including an artifact-correcting process;

FIGURE 3A diagrammatically shows an uncorrected reconstructed image;

FIGURE 3B diagrammatically shows a classified image; and

FIGURE 3C diagrammatically shows a metal-free image.

With reference to FIGURE 1, a imaging system 10 includes a computed tomography scanner 12 having a radiation source 14 that produces a radiation beam

directed into an examination region **16**. The radiation beam interacts with and is partially absorbed as it traverses a region of interest of an imaging subject disposed in the examination region **16**, producing spatially varying absorption of the radiation as it passes through the examination region. A radiation detector **18** detects the absorption-attenuated radiation after it passes through the examination region **16**.

Preferably, the radiation source **14** produces a fan-beam or cone-beam of x-rays. The radiation source **14** and the detector **18** are preferably mounted in oppositely facing fashion on a rotating gantry **20** so that the detector **18** continuously receives x-rays from the radiation source **14**. As the source **14** and the detector **18** rotate about the examination region **16** on the rotating gantry **20**, views are acquired over an angular range of preferably about 360° or more. Optionally, a reduced scan of between about 180° and 360° is used. In one embodiment, the detector **18** is replaced by a stationary detector ring mounted on a stationary gantry **22**. Typically, a subject support **26** is linearly movable in an axial or z-direction by a motor means **28**.

Multiple-slice computed tomography projection data are acquired by performing successive axial scans with the subject support **26** being stationary during each axial scan and stepped linearly between axial scans. In this arrangement, the detector **18** can have either a single row of detector elements (that is, a one-dimensional detector) or a two-dimensional array of detector elements. Alternatively, helical computed tomography projection data are acquired during continuous linear movement of the subject support **26** and simultaneous rotation of the gantry **20**.

The outputs of detector elements of the radiation detector **18** are converted to electric acquired integrated attenuation projection values μd_o that are stored in a data memory **30**. Each projection datum μd_o corresponds to a line integral of attenuation along a line from the radiation source **14** to a corresponding one of the detector elements of the detector **18**. The projection data can be represented in a sinogram format in which each two-dimensional slice of the imaged region of interest is represented by a projection data array having coordinates of viewing angle (φ) and line integral index (n).

For typical fan-beam and cone-beam geometries, the line integral index n typically corresponds to a detector index indicating a detector element used to measure the projection of index n. It is contemplated, however, that the line integral index n may lack a

direct correspondence with detector element number. Such a lack of direct correspondence can result, for example, from interpolation between rebinned projections.

With continuing reference to FIGURE 1, a slice cycling processor **32** cycles through the sinograms corresponding to spatial slices and successively inputs each sinogram into a reconstruction processor **34**. The reconstruction processor **34** reconstructs the input data into a 3D uncorrected tomographic image **T** which is stored in an uncorrected 3D image memory **36**. Although picture elements of a 2D or slice image are commonly denoted as "pixels", and elements of a 3D or volume image are often denoted as "voxels"; "pixels" is used herein generally to refer to picture elements of both 2D and 3D images.

With reference to FIGURE 3A, the presence of one or more high density regions **40** in the slice typically causes the uncorrected reconstructed image **T** to include metal artifacts which generally manifest in the image as streaks **42** extending away from the high density region **40**. The diagrammatic reconstructed image **T** of FIGURE 3A is not intended to correspond to images of any particular anatomical structure, but rather diagrammatically illustrate features of the artifact-correcting reconstruction process that are typically observable in image space.

With reference again to FIGURE 1, a correction means or processor or algorithm **44** performs artifacts correction, in which artifacts introduced by high density regions such as metal clips, high-density dental fillings, or the like, are substantially corrected, even for images that contain discontinuous high density regions or one or more medium density regions in addition to the one or more high density regions. As described in a greater detail below, the correction processor **44** receives an uncorrected image data and outputs a corrected image data which is consequently reconstructed by the reconstruction processor **34** into a corrected 3D tomographic image representation **T'**. Spatially successive artifact-corrected reconstructed image slices, slabs or volumes are accumulated in a corrected 3D image memory **46** to define a three-dimensional artifact-corrected reconstructed volume image. If, however, the acquired projection data is limited to a single slice of the region of interest, then the acquired projection data corresponding to the single slice is processed by the reconstruction processor **34** and the corrected 3D image memory **46** stores a two-dimensional artifact-corrected reconstructed image. Optionally, projection data corresponding to one or more image slices are acquired

over a selected time interval to provide a temporal series of artifact-corrected reconstructed image slices or image volumes representative of a temporal evolution of the region of interest.

A video processor **50** processes some or all of the contents of the corrected 3D image memory **46** or, optionally, of the uncorrected 3D image memory **36** to create a human-viewable image representation such as a three-dimensional rendering, a selected image slice, a maximum intensity projection, a CINE animation, or the like. The human-viewable image representation is displayed on a display **52** of a user interface **54**, which is preferably a personal computer, a workstation, a laptop computer, or the like. Optionally, selected contents of image memory **36, 46** are printed on paper, stored in a non-volatile electronic or magnetic storage medium, transmitted over a local area network or the Internet, or otherwise processed. Preferably, a radiologist or other operator controls the computed tomography imaging scanner **12** via an input means **56** to program a scan controller **58** to set up an imaging session, modify an imaging session, execute an imaging session, monitor an imaging session, or otherwise operate the scanner **12**.

With continuing reference to FIGURE 1 and further reference to FIGURE 2, the correction means **44** receives the filtered uncorrected tomographic image data **T** and applies a clustering technique or process **60** to each pixel in the image data set. The data is clustered into classes so that the data within each class is more similar than those outside that class or in another class. In one embodiment, in which the original sinogram data **S** is not available, the uncorrected tomographic image **T** is initially filtered by a tomogram filter means **62**, which applies a low pass filter to harmonize noise. Preferably, the filter means **62** applies a two-dimensional Gaussian filter with a 5 by 5 pixel mask and a standard deviation (sigma) of 1. The filtered uncorrected tomographic image **T** is received as the input by the correction means **44** to compensate for metal artifacts.

With continuing reference to FIGURE 2 and further reference to FIGURE 3B, the clustering means **60** partitions the uncorrected tomographic image data **T** into the clusters using a statistical classifier. Preferably, the clustering means **60** employs a statistical k-means classifier. More particularly, a k-means **64** supplies a number of classes and each class properties to the clustering means **60**. Preferably, at least four default classes are identified in advance by the user: metal class "4", bones class "3", soft tissue class "2", and air class "1". Alternatively, fixed sets of settings are identified for particular

applications and used automatically. The initial mean value for each class is autogenerated and preassigned. Optionally, each class is assigned an equal number of pixels. Optionally, the classes and initial values are defined interactively by the user. The user interactively intervenes, when required, to redefine and reassign the classes and initial values. A

5   segmentation or classifying means or algorithm **66** classifies pixels in the uncorrected tomographic image **T** into the identified classes to generate a segmented or classified image T_{class} in which pixel values are replaced by class classification index values corresponding to the metal, bone, soft tissue, and air classes.

With continuing reference to FIGURE 3B, the pixel classified image T_{class},

10   which is diagrammatically shown in FIGURE 3B, has regions of the image essentially consisting of pixels of a particular class labeled by an appropriate class index selected from the class indices "4", "3", "2", "1". The exemplary image includes a region of high density or metal class "4" which could be a metal implant, a region of medium density or bone class "3" which could be a region of bone, both contained within a region of low density or

15   tissue class "2" which could be soft tissue or the like. A region of air density or air class "1" fills the periphery of the pixel density-classified image T_{class} corresponding, for example, to the ambient air surrounding the imaging subject.

With reference again to FIGURE 2, the clustering means **60** determines the centroid of each cluster. The clustering means **60** utilizes an iterative algorithm which

20   minimizes the sum over all clusters of within-cluster sums of pixel value -to-cluster centroid distance. As a result, some of the pixels are reassigned to different clusters. New centroids are determined as the input to minimize. The clustering process **60** is continued until the optimum assignment of all pixels is found. The result is a vector containing the cluster indices of each pixel and a vector containing the cluster values. The classified pixel

25   data are stored in a classified pixel memory **68**. It is contemplated that other statistical classifiers such as c-mean, fuzzy c-mean, unsupervised Bayesian, and the like might be used.

A metal pixels replacement means **70** assigns a surrounding tissue class value to the pixels or voxels classified into the metal class. Preferably, the bone class value

30   is predetermined as a default value for such pixels. Of course, it is also contemplated that the tissue class value or other class value might be assigned to such pixels. The finalized classified pixels are stored in a final classified pixel memory **72**.

With reference to FIGURE 3C, in the final classified image T_{class_final} the regions of high density "4" are effectively removed by replacing pixels in these regions with the grayscale value for pixels of the medium density class "3", e.g. bone.

With reference again to FIGURE 2, although not shown in FIGURE 3C, one or more of the regions may have discontinuances, sharp points, or other anatonomically incorrect shapes. A morphological means or algorithm 80 performs morphological operations on the image to generate a tomogram model image T_{model}. Optionally, the user selects selected regions, e.g. the regions containing severe artifacts, to perform morphological operations on. Preferably, the morphological means 80 uses opening and/or closing algorithms or operators 82, 84 to restore the image content. The opening and closing algorithms 82, 84 use prior knowledge which is inserted into the image to eliminate unwanted artifacts which are known not to be present in the soft tissue by the virtue of human anatomy. For example, it is known that there is no sharp edges in the soft tissue; thus, prior knowledge that there is no sharp edges is incorporated. Preferably, the morphological means 80 eliminates air bubbles, narrow spikes, and other inappropriate data that is known not to be present in the soft issue. Preferably, the shape of the opening/closing operators 82, 84 is a disk with a size between five to eight pixels. Preferably, the user sets the default size and shape, e.g. a disk of six pixels wide, ahead of time. In one embodiment, the user interactively modifies the default size and shape depending on the artifact level: the stronger the artifacts are, the bigger the size of the operator 82, 84 is set at. The resultant tomogram model image T_{model} is stored in a tomogram model memory 86. A forward projection means 88 forward projects the tomogram model image T_{model} to produce a sinogram model image S_{model} which is stored in a sinogram model image memory 90.

A corrupted rays identifying means 100 identifies rays or class "4" regions in the original sinogram data S that pass through the metal, i.g. the corrupted rays to be corrected. Preferably, the corrupted rays identifying means 100 identifies edges/boundaries of regions consisting of the pixels of the high density by one of known techniques after the metal classified pixels are forward projected onto a sinogram..

A corrupted rays replacement means 102 replaces the identified corrupted rays of the original sinogram data S, e.g. projection data corresponding to the rays passing through high density regions, with the projection data from the sinogram model S_{model} to

produce a corrected sinogram image S'. Preferably, the corrupted rays replacement means 102 calculates a linear offset or other smooth transition that enables a smooth integration of the corresponding data portions of the sinogram model into the original sinogram. In this manner, the corrupted regions of the original sinogram image S are replaced by the

5   corresponding regions from the sinogram model S_{model}.

The metal-free, corrected sinogram image S' is stored in a corrected sinogram image memory 104.

With reference again to FIGURE 1, the reconstruction processor 34 uses a standard filtered backprojection means or algorithm 110 to generate a corrected

10   tomographic image. The backprojection means 110 backprojects the corrected sinogram S' to obtain corresponding intermittent corrected tomographic image T'_{int}. Preferably, the reconstruction processor 34 uses a Ramachandran-Lakshminarayanan filter. To get a clearer view of the corrections, an extraction means 112 extracts the metal segments from the original uncorrected tomographic image T. A combiner 114 superimposes or combines

15 .   the extracted metal segments with the intermittent corrected tomographic image T'_{int} to generate a final corrected tomographic image T' which is stored in the corrected 3D image memory 46.

Preferably, the correction process 44 is iteratively repeated such that the classes are redefined and modified. For example, on a subsequent iteration the k-means

20   64 supplies five classes instead of four classes to further redefine and differentiate data.

The invention has been described with reference to the preferred embodiments. Obviously, modifications and alterations will occur to others upon reading and understanding the preceding detailed description. It is intended that the invention be construed as including all such modifications and alterations insofar as they come within

25   the scope of the appended claims or the equivalents thereof.